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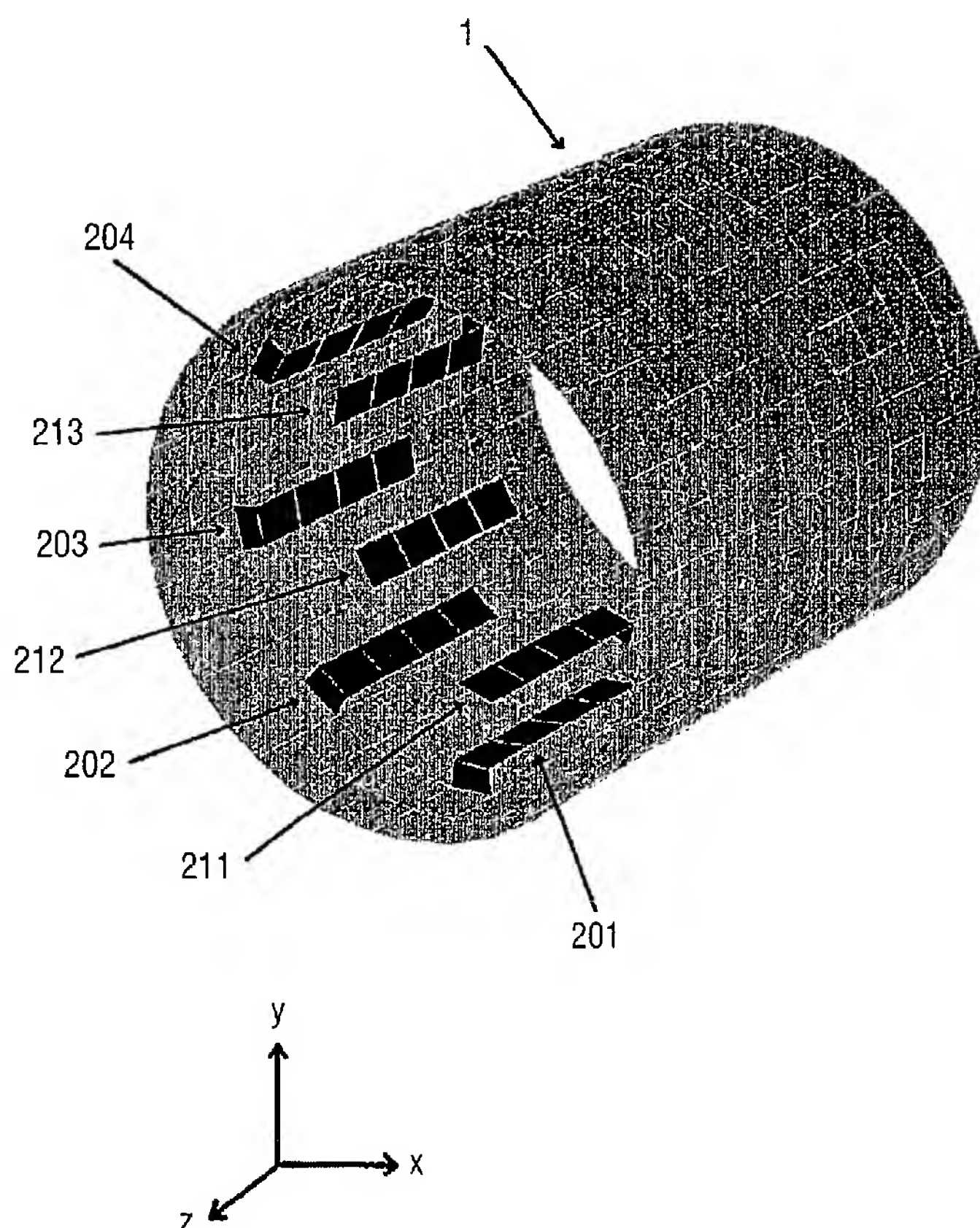
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LK, LR, LS, LT, LU, LV, MA, MD, MG, MK, MN, MW,
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(54) Title: RF COIL SYSTEM FOR A MAGNETIC RESONANCE IMAGING APPARATUS



(57) Abstract: The invention relates to a device for the transmission and/or reception of RF signals for magnetic resonance imaging (RF coil system) which is constructed as an RF coil (body coil) which is permanently mounted in a magnetic resonance imaging apparatus or as a so-called dedicated RF coil (that is, as a separate RF coil which is to be arranged on or around a region to be examined, for example, a head coil, a shoulder coil, a flexible surface coil etc.), as well as to a magnetic resonance imaging apparatus which comprises an RF coil system of this kind. The RF coil system is characterized in that it consists of a plurality of resonant conductor elements (201, 202, ..., 208; 211, 212, ..., 218), notably $\lambda/4$ monopole elements. In comparison with known RF coil systems of the same length, a substantially larger field of view as well as a constant variation and a steeper drop of the field characteristic are thus achieved at the edges of the RF coil system while the construction remains comparatively simple and economical nevertheless.



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RF coil system for a magnetic resonance imaging apparatus

The invention relates to a device for the transmission and/or reception of RF signals for magnetic resonance imaging (referred hereinafter in general as an "RF coil system") which is constructed as an RF coil which is permanently mounted in a magnetic resonance imaging apparatus (body coil) or as a so-called dedicated RF coil (that is, as a
5 separate RF coil which is to be arranged on or around a region to be examined, for example, a head coil, a shoulder coil, a flexible surface coil etc.), as well as to a magnetic resonance imaging apparatus provided with such an RF coil system.

Magnetic resonance (MR) imaging apparatus is used notably for the examination and treatment of patients. The nuclear spins of the tissue to be examined, aligned
10 by a steady main magnetic field (B_0 field) are then excited by a pulse-like B_1 field which is orthogonal to the main magnetic field and has the MR or Larmor frequency. Moreover, for localization the nuclear spins are also subjected to gradient magnetic fields. The RF relaxation signals induced by the excitation are received and evaluated in order to reconstruct an image of the relevant tissue therefrom in known manner.

15 Essentially two types of construction can be distinguished: on the one hand there are the so-called axial systems in which the patient is introduced into an essentially horizontally oriented tubular examination zone. The magnetic fields are generated by magnet coils which are arranged along the circumference of the examination zone, the main magnetic field then traversing the patient in the direction of the longitudinal axis of the patient.

20 On the other hand, there are the so-called open MR imaging apparatus (vertical systems) in which the main magnetic field is generated, generally speaking, between two pole plates which are arranged one above the other and wherebetween a vertical cylindrical examination zone for a patient is defined. The main magnetic field (B_0 field) traverses the patient essentially in a direction perpendicular to the longitudinal axis of the
25 patient (that is, vertically). The patient then remains suitably accessible from almost all sides, that is, even during the imaging, so that interventional examinations can also be performed.

RF coil systems are permanently mounted in said systems (so-called RF body coils) in order to generate the B_1 field (RF field) and to receive the RF relaxation signals; the

configuration and positioning of such coils has a decisive effect on the image quality, notably the signal-to-noise ratio and the resolution.

Moreover, dedicated (separate) RF coils such as, for example, head coils, shoulder coils etc., are also used; these coils are also known as at least partly flexible surface
5 coils or pads and can be arranged around or on the region of a patient to be examined.

In this respect it is very important that the overall examination zone to be imaged is traversed by an as homogeneous as possible RF field or that the reception characteristic of the RF coil system is as constant as possible in this zone. Furthermore, the field of view (FOV) of the RF coil system should extend as accurately as possible across the
10 space of the at least substantially constant B_0 field and the useful space of the gradient coils, generating the gradient magnetic fields, so that the resultant, defined useful examination zone (and only this zone) is exposed as completely and constantly as possible so as to be used for disturbance-free imaging.

15

For example, United States patent 6,150,816 discloses an RF coil system which consists of at least a first, a second and a third RF coil which are electrically insulated from one another, can be separately activated and are arranged so as to overlap one another in the axial direction in such a manner that no magnetic coupling exists therebetween. The aim
20 is to provide not only an improved signal-to-noise ratio but also an expanded and switchable field of view of the RF coil system.

However, this system has the drawback that the length of this coil system in the axial direction is comparatively large in comparison with the length of the field of view in this direction, that is, in particular when an as constant as possible variation and a steep drop
25 of the RF field or the reception characteristic are desired at the axial ends of the field of view.

Therefore, it is a general object of the invention to provide a device for generating B_1 fields and/or for receiving RF relaxation signals (customarily referred to as an
30 RF coil system) whereby a useful examination zone can be exposed with a field of view which is constant to a high degree in respect of the transmission or reception characteristic (field characteristic).

It is notably an object of the invention to provide an RF coil system of the kind set forth for an MR imaging apparatus in the form of an axial system whereby a field

characteristic can be generated which is adapted to the dimension of the at least substantially constant B_0 field in the axial direction and to the gradient magnetic fields and which decreases comparatively steeply at the axial ends while utilizing a comparatively small dimension of the RF coil system in this direction.

5 Finally, it is also an object of the invention to provide an RF coil system of the kind set forth for an MR imaging apparatus in the form of a vertical system whereby a field characteristic can be obtained which is adapted to the dimension of the at least substantially constant B_0 field in the radial direction of the examination zone and to the gradient magnetic fields and which decreases comparatively steeply at the radial ends.

10 The object is achieved in conformity with claim 1 by means of a device for the transmission and/or reception of RF signals for magnetic resonance imaging (RF coil system) which is formed by a plurality of resonant conductor elements.

 This RF coil system thus constitutes essentially a wave resonator. A special advantage of this solution resides in the fact that in an axial system the ratio of the dimension
15 of the field of view in the direction of the axis of the RF coil system to the axial length of this RF coil system can thus be increased by approximately the factor 2.

 Moreover, according to this solution not only RF coils which are permanently mounted in the system can be realized, but also the previously mentioned dedicated RF coils.

 A further advantage of this solution consists in the fact that in comparison with
20 known RF coil systems it can be constructed in a comparatively simple and economical manner.

 The dependent claims relate to advantageous further embodiment of the invention.

 The embodiments disclosed in the claims 2 and 3 enable a desired
25 transmission or reception characteristic to be achieved in a comparatively simple and economical manner.

 The embodiment disclosed in claim 4 offers the advantage that the tuning to the MR frequency of a tissue to be examined can be performed in a simple manner.

 The claims 5 and 6 describe an embodiment of the RF coil system which is
30 intended to be mounted permanently in an axial system or is intended for the use as an RF volume coil, whereas the embodiment disclosed in claim 7 is intended for a vertical system or an RF surface coil.

Further details, characteristics and advantages of the invention will become apparent from the following description of preferred embodiments which is given with reference to the drawing. Therein:

Fig. 1 is a diagrammatic three-dimensional representation of a body coil with a first RF coil system in accordance with the invention;

Fig. 2 is a plan view of the circumference of the body coil from the inside;

Fig. 3 is a cross-sectional view of the body coil;

Fig. 4 shows a variation of the magnetic field strength of a conductor over a plane;

Fig. 5 shows a variation of the magnetic field strength of a conductor over a plane;

Fig. 6 shows a typical variation of the magnetic field strength in the first RF coil system, and

Fig. 7 is a plan view of a second RF coil system for an MR imaging apparatus in the form of a vertical system.

Fig. 1 is a diagrammatic representation of a quadrature body coil (QBC) which comprises an external RF shield 1 of an electrically conductive material.

On the inner circumference of the RF shield 1 there is provided a first embodiment of an RF coil system in accordance with the invention. This RF coil system comprises a first group of conductor elements 201, 202, 203, ... which are distributed along the circumference as well as a second group of conductor elements 211, 212, 213, ... which are also distributed along the circumference. All conductor elements extend in the axial direction of the examination zone (z direction) and have the same length. The conductor elements of a group are situated essentially at the same level in the z direction.

In order to realize a given field characteristic, however, the conductor elements may also have a different length and/or be situated at a different level in the z direction.

The two groups of conductor elements are arranged so as to be offset relative to one another in the z direction, so that, viewed in the circumferential direction of the examination zone, the conductor elements partly overlap one another and hence are coupled to one another. Viewed in the circumferential direction, the conductor elements 201, 202, ...; 211, 212, ... then belong alternately to the first group and the second group. Granted, a

coupling between the conductors also occurs in the absence of mutual overlapping, but in that case the field characteristic possibly is not as uniform as it is now.

The electrically effective length of the circumference on which the conductor elements 201, 202, ...; 211, 212, ... are situated, which length is determined by the dielectric constant of the surrounding materials (for example, the tuning structure and the patient shielding as described hereinafter), is proportioned so that it corresponds approximately to the wavelength λ of the RF frequency (Larmor frequency).

For the use of the RF coil system at low frequencies, furthermore, a further dielectric structure may be added so as to reduce its electrical length for a given circumference.

The length of the conductor elements 201, 202, ...; 211, 212, ... is proportioned to be such that these elements are resonant at the RF frequency and hence, as opposed to known RF coil systems, a non-constant current distribution occurs along the conductor elements 201, 202, ...; 211, 212,

Each of the conductor elements 201, 202, ...; 211, 212, ... is formed by a resonant conductor and preferably a monopole element of a metallic foil having a length of $\lambda/4$. The ends of the monopole elements which are situated at the area of overlap of the two groups are open, whereas the oppositely situated (that is, the axially outer) ends of the monopole elements are coupled to the RF shield 1.

The axially inner ends of the monopole elements can also be connected to the RF shield 1 via capacitors in order to reduce their electrical length, so that the current does not drop to the value zero in these regions. The variation of the field characteristic of the RF coil system in these regions can thus be more suitably adapted to the variation in the axially outer regions.

Fig. 2 is a plan view of the RF shield 1 in an unrolled state, that is, from within the examination zone. The two partly overlapping groups 20, 21 of monopole elements 201, 202, ..., 208 and 211, 212, ..., 218 can again be recognized in this representation.

Furthermore, in conformity with this representation a tuning structure in the form of a strip-like element 22, 23 (which is closed so as to form a ring in the rolled-up state) of a dielectric material is provided for each group 20, 21; this tuning structure extends essentially perpendicularly to the longitudinal direction of the monopole elements and has a periodically recurrent increased width and/or thickness in conformity with the number of monopole elements 201, 202, 203, ..., 208 or 211, 212, 213, ..., 218 in each group 20, 21. In

the case shown one side of the element 22, 23 (or of the ring) has the shape of a sawtooth for this purpose.

Furthermore, the elements or rings 22, 23 are arranged in such a manner that (together or independently from one another) they can be displaced (or rotated) in the circumferential direction of the RF shield 1 (in this case in conformity with the arrow A) in such a manner that, in dependence on the position, a segment of the ring 22 or 23 with a smaller or larger width will be situated in the region of each monopole element 201, 202, 203, ..., 208 or 211, 212, 213, ..., 218, thus modifying the characteristic impedance of parts of the monopole elements.

The RF coil system can thus be tuned to the Larmor frequency of the tissue examined by rotating the dielectric rings 22, 23 in the circumferential direction.

Fig. 3 is a cross-sectional view (x/y plane) of the RF coil system shown in Fig. 1. Figure 3 shows the RF shield 1 as well as the resonant monopole elements 201, 202, ..., 208 of the first group 20. The rings 22, 23 of a dielectric material (tuning structures) are not shown.

Figure 3 also shows a number of diodes (for example, pin diodes) D1, D2, ..., D8 which are coupled to the RF shield 1 each time via the outer ends of the monopole elements 201, 202, ..., 208 in the axial direction of the examination zone. The oppositely situated free ends of the monopole elements 201, 202, ..., 208 are connected to one another via inductances, for example, choke coils (not shown) which constitute a very high impedance for RF signals.

The diodes D1, D2, ..., D8 can thus be rendered conductive or be blocked by application of a bias voltage between the RF shield 1 and the free ends of the monopole elements 201, 202, ..., 208 which are connected via the choke coils. In the conductive state the monopole elements 201, 202, ..., 208 are connected to the RF shield 1 via the diodes D1, D2, ..., D8, so that the RF coil system operates in the described manner. When the diodes D1, D2, ... are blocked by an appropriate change of the bias voltage, the RF coil system is detuned. This makes sense, for example, when different RF coil systems are used for the transmission and the reception and when the RF coil system which is inactive at the moment is to be prevented from disturbing the active RF coil system by resonance or coupling effects.

Finally, in Fig. 3 a further RF shield 24 (patient shield) which surrounds the patient is provided in the examination zone. This prevents the patient from being exposed to the comparatively high electrical field which occurs at the free ends of the monopole elements, so that a high specific absorption rate (SAR) arises.

The shield 24 can also serve as a tuning structure for the RF coil system if it is provided with a periodically recurrent change of cross-section, as described above with reference to the tuning structures 22, 23, and when it is arranged so as to be rotatable.

As has already been stated, the described RF coil system ensures that the
5 current distribution along the conductor elements is not constant but various in a cosine-like fashion because of the resonant length at the RF frequency.

Figs. 4 and 5 illustrate the resultant effect on the variation of the magnetic field strength H. Fig. 4 shows the variation occurring in the case of a conductor L with a constant current distribution which is arranged over a conductive plane E, whereas Fig. 5
10 shows the variation in the case of a cosine-like current distribution in such a conductor L.

In the case of $\lambda/4$ monopole elements, the current in the monopole elements has a maximum at the ends which are situated in the axial direction and are coupled to the RF shield 1 whereas it drops to a minimum or zero in an essentially cosine-like fashion in the direction of the inner open ends.

15 The monopole elements of each group 20, 21 are electrically and magnetically coupled to one another and have resonances in different modes of operation. These modes of operation enable a very constant magnetic field to be generated in a plane perpendicular to the axis of the examination zone (x/y plane). These modes of operation are split each time into an even mode and an odd mode of operation as a result of the arrangement of two groups
20 of monopole elements.

In the odd (low) modes of operation the currents flowing in the z direction in neighboring monopole elements have the same phase and exhibit a maximum each time at the (axial) ends situated in the z direction of the RF coil system, so that overall a substantially constant magnetic field variation is obtained in the z direction and the field of view in this
25 direction is significantly larger in comparison with known RF coil systems.

An even steeper drop, and hence a substantially more rectangular variation of the field characteristic in the z direction, can be achieved by the addition of further groups of monopole elements (not shown) at the two axial ends of the RF coil system, that is, if the currents in the monopole elements of the further groups exhibit a phase shift of 180 degrees
30 relative to the currents in the neighboring monopole elements of the first or the second group 20, 21.

Fig. 6 shows diagrammatically a variation of the magnetic field strength H_y , produced by the first RF coil system, within the RF shield 1 in the x/z (or x/z) plane, two of the monopole elements 201, 205 also being shown. In comparison with a known birdcage

coil (for example, a head coil), the ratio of the field of view to the coil length can be increased, for example, by approximately the factor 2.

The principle in accordance with the invention can also be used for the previously described open MR imaging apparatus (vertical systems). Fig. 7 shows a second
5 embodiment of the invention in the form of an RF coil system which is rigidly arranged as a transmission and/or receiving antenna at the area of at least one axial end of the vertical-cylindrical examination zone of the MR imaging apparatus.

This RF coil system is also provided with an RF shield 1a which again serves as a carrier for a group of resonant conductor elements 301, 302, ..., 312. The conductor
10 elements are arranged so as to extend radially from the center of the RF shield 1a. The radial outer ends of the conductor elements 301, 302, ..., 312 are coupled to the RF shield 1a also in this case.

This coupling can again be realized via a respective diode (not shown), the oppositely situated radial inner ends then being connected to one another via inductances
15 which constitute a very high impedance for the RF signals. If necessary, the RF coil system, can be detuned by application of a bias voltage in conformity with the foregoing description.

Furthermore, in conformity with the foregoing description the radial inner ends of the conductor elements 301, 302, ..., 312 are either open or, as indicated in Fig. 7, connected to the RF shield 1a via capacitors C1, C2, ..., C12 in order to reduce their electrical
20 length so that at these areas the current does not drop to the value zero. A field variation can thus be achieved which is more constant in the radial direction.

In this embodiment the conductor elements 301, 302, ..., 312 are preferably $\lambda/4$ monopole elements of a metallic foil again, which elements may become wider in the radially outwards direction. Furthermore, the previous explanations given in relation to the
25 first embodiment of the RF coil system also hold for the operation of this RF coil system.

Furthermore, this embodiment can again be provided with a tuning structure in conformity with the foregoing description, for example, in the form of one or more concentric rings which have a periodically varying width and/or thickness along the circumference and consist of a dielectric material.

30 In both embodiments the conductor elements may be connected to one another via capacitors which are arranged along the length of the conductor elements in order to modify their coupling and hence adapt the field characteristic of the RF coil system to given requirements.

For the sake of completeness it is to be noted that, in order to generate a circularly polarized magnetic field, in the RF coil systems in accordance with the invention the power supply to a plurality of conductor elements can also take place with suitably phase-shifted RF signals or the received RF signals can be coupled out in a corresponding fashion
5 via such conductor elements.

CLAIMS:

1. A device for the transmission and/or reception of RF signals for magnetic resonance imaging (RF coil system) which is formed by a plurality of resonant conductor elements (201, 202, ..., 208; 211, 212, ..., 218; 301, 302, ..., 312).
- 5 2. An RF coil system as claimed in claim 1, in which the conductor elements are RF-coupled to an RF shield (1, 1a) by way of one end and while their other end is open.
3. An RF coil system as claimed in claim 1, in which the conductor elements are $\lambda/4$ monopole elements (201, 202, ..., 208; 211, 212, ..., 218; 301, 302, ..., 312).
- 10 4. An RF coil system as claimed in claim 1, comprising at least one tuning element (22, 23) of a dielectric material which is arranged so as to be displaceable essentially perpendicularly to the longitudinal direction of the conductor elements (201, 202, ..., 208; 211, 212, ..., 218; 301, 302, ..., 312) and has a segment of different width and/or thickness at
15 the area of the conductor elements, so that the resonance frequency of the RF coil system can be tuned by displacement of the tuning element (22, 23).
5. An RF coil system as claimed in claim 1, in which the conductor elements (201, 202, ..., 208; 211, 212, ..., 218) are arranged on the inner surface of an essentially
20 cylindrical RF shield (1) while their longitudinal direction extends in the axial direction of an examination zone enclosed thereby.
6. An RF coil system as claimed in claim 5, in which the conductor elements (201, 202, ..., 208; 211, 212, ..., 218) are arranged in a first group and a second group (20, 21), the two groups (20, 21) being arranged so as to be offset relative to one another in the
25 axial direction of the examination zone and the conductor elements (201, 202, ..., 208; 211, 212, ..., 218) being coupled to the RF shield (1) by way of their outer ends in the axial direction.

7. An RF coil system as claimed in claim 1, in which the conductor elements (301, 302, ..., 312) are arranged on an essentially flat RF shield (1a), their length dimension extending in radial directions from a center.

5 8. A dedicated RF coil for use in magnetic resonance imaging, comprising an RF coil system as claimed in claim 1.

9. A magnetic resonance imaging apparatus comprising a horizontal-cylindrical examination zone (axial system) as well as an RF coil system as claimed in claim 5.

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10. A magnetic resonance imaging apparatus, comprising a vertical-cylindrical examination zone (vertical system) as well as an RF coil system as claimed in claim 7.

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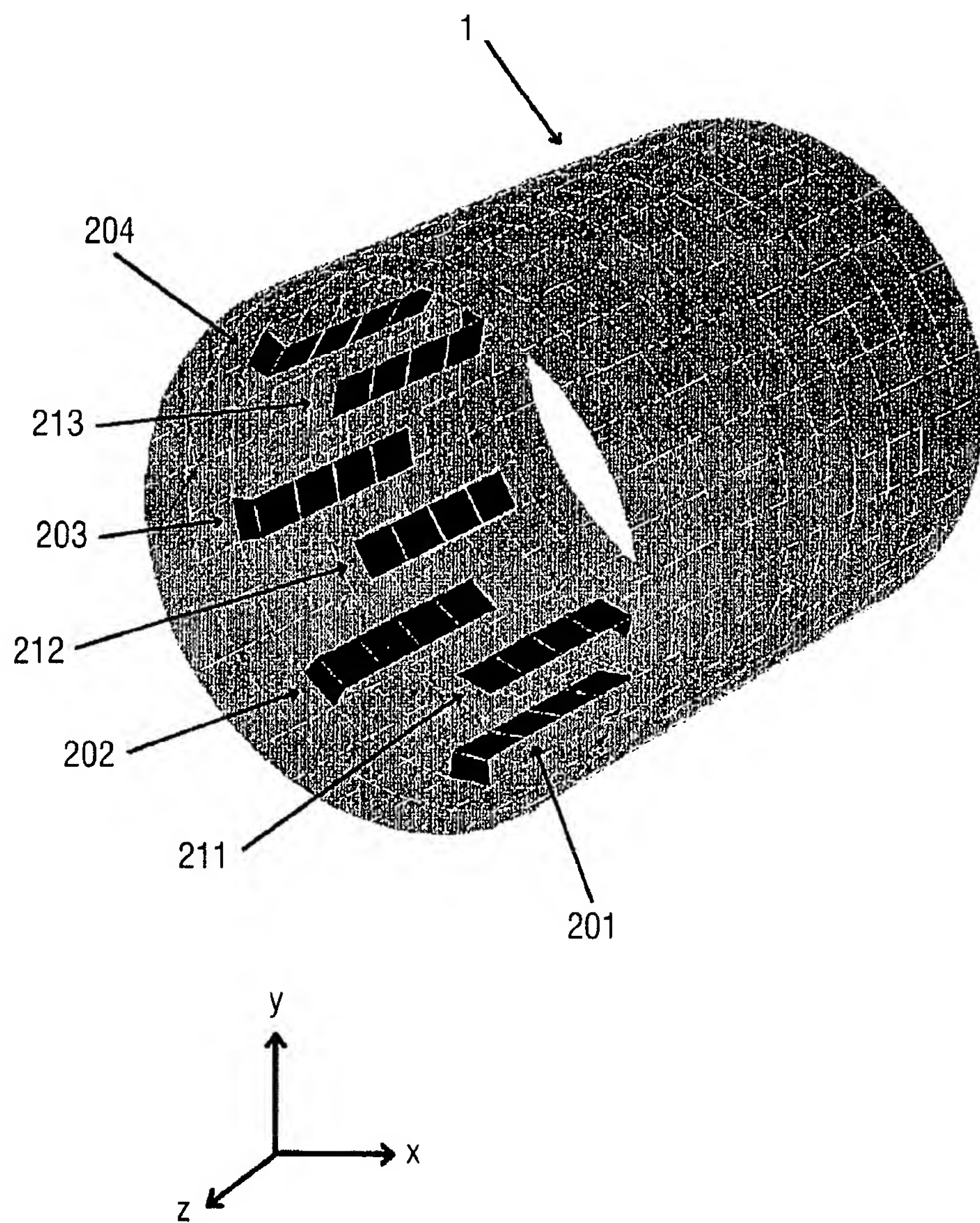


FIG. 1

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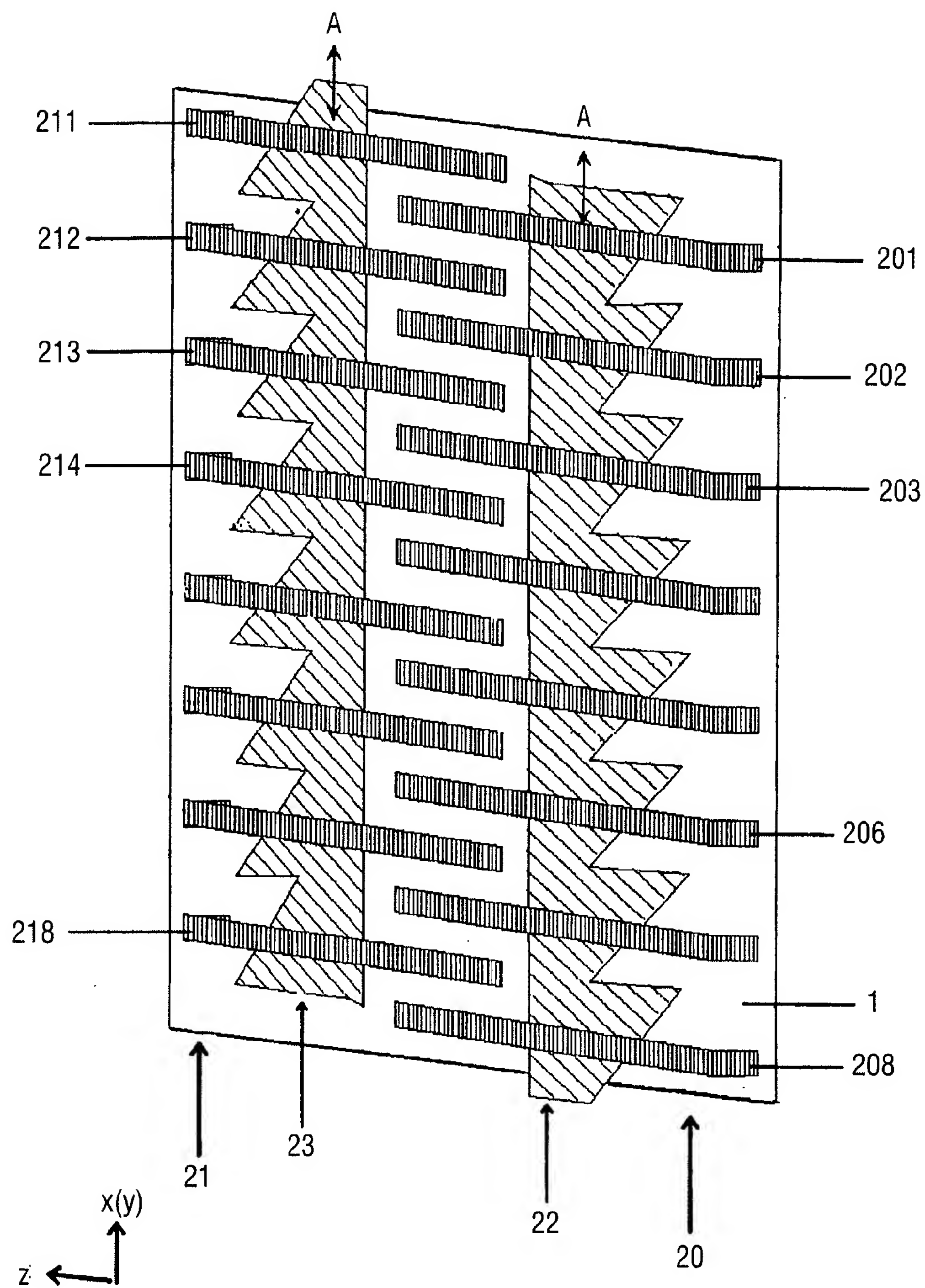


FIG. 2

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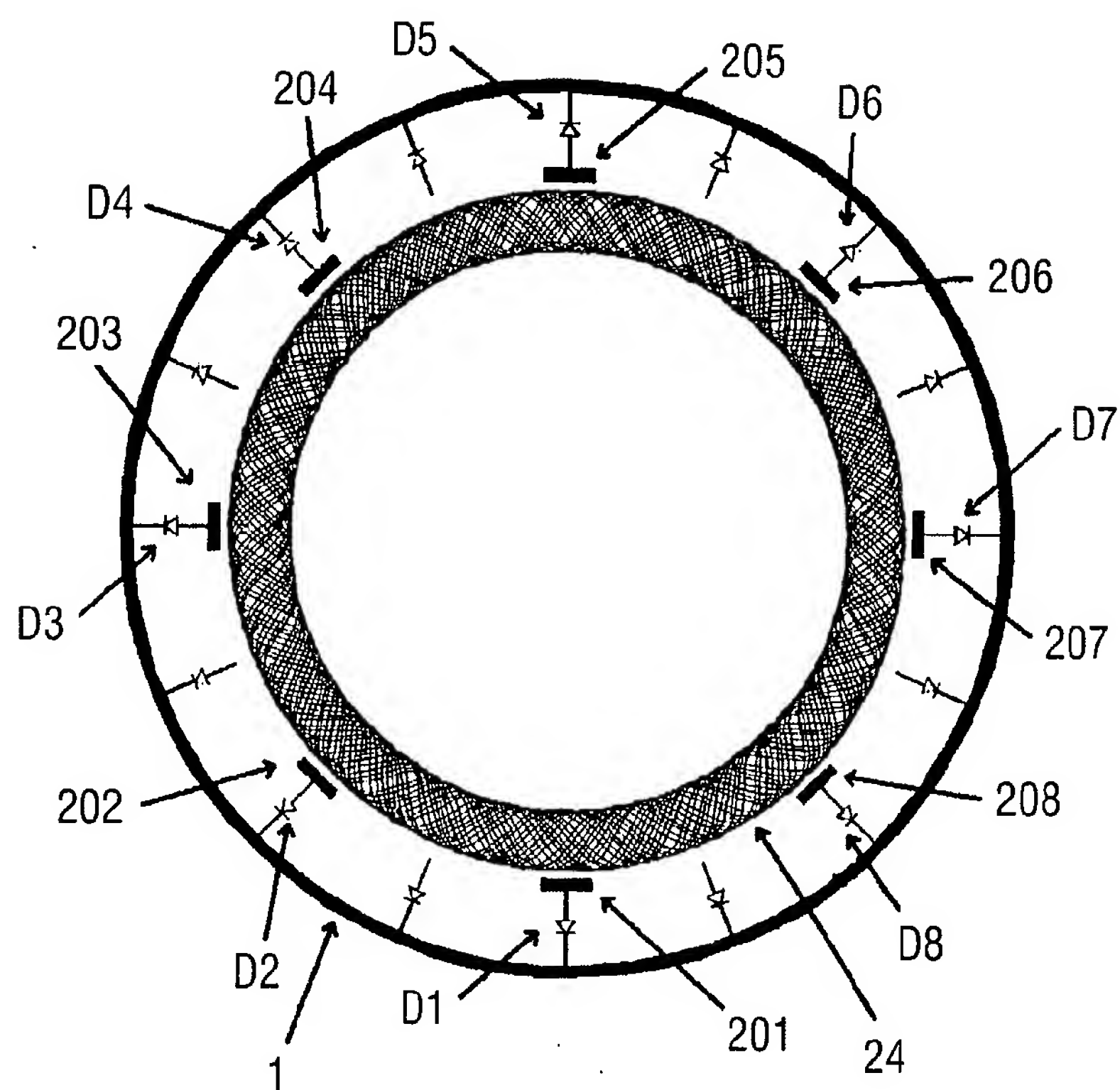
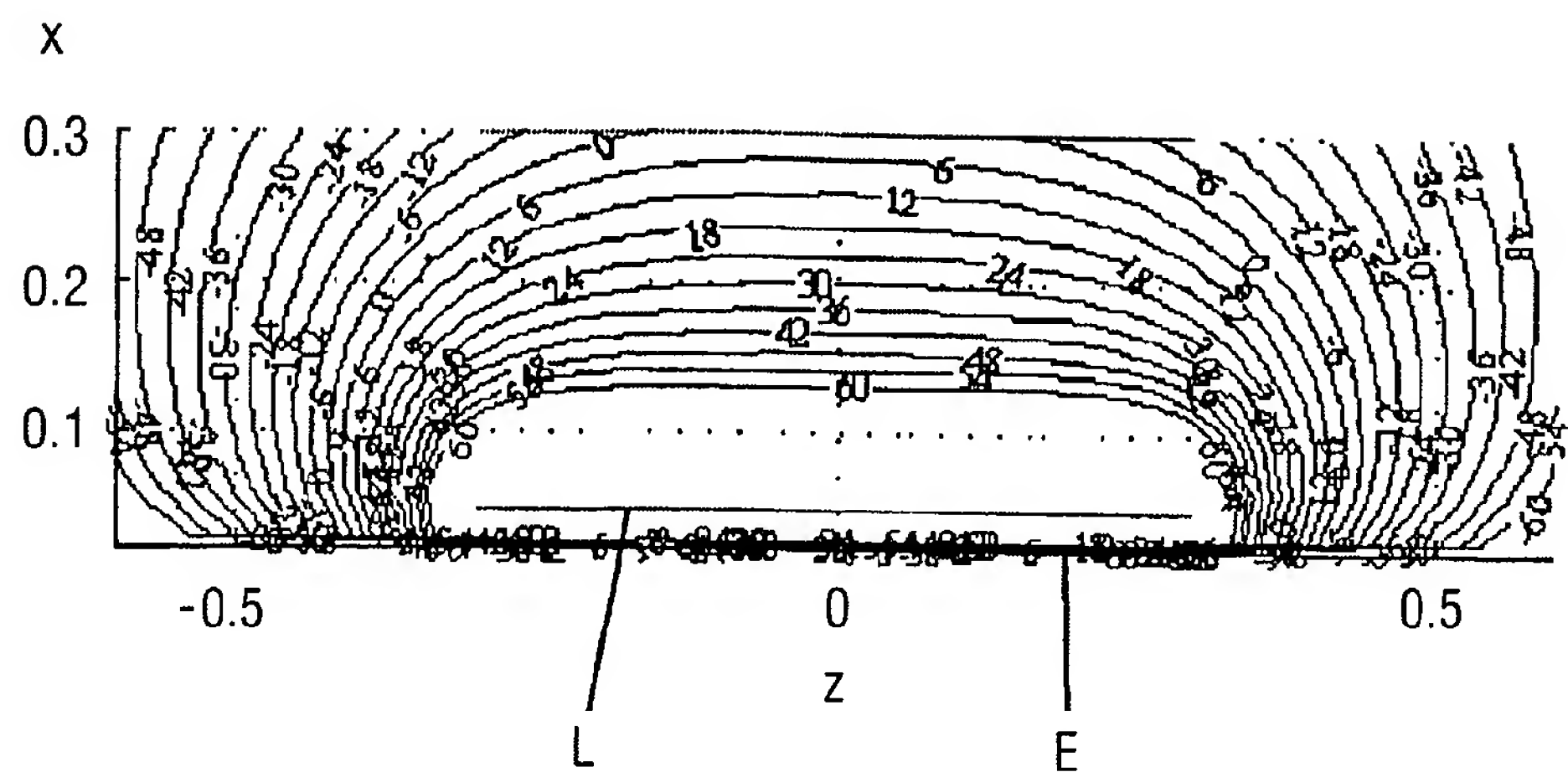


FIG.3



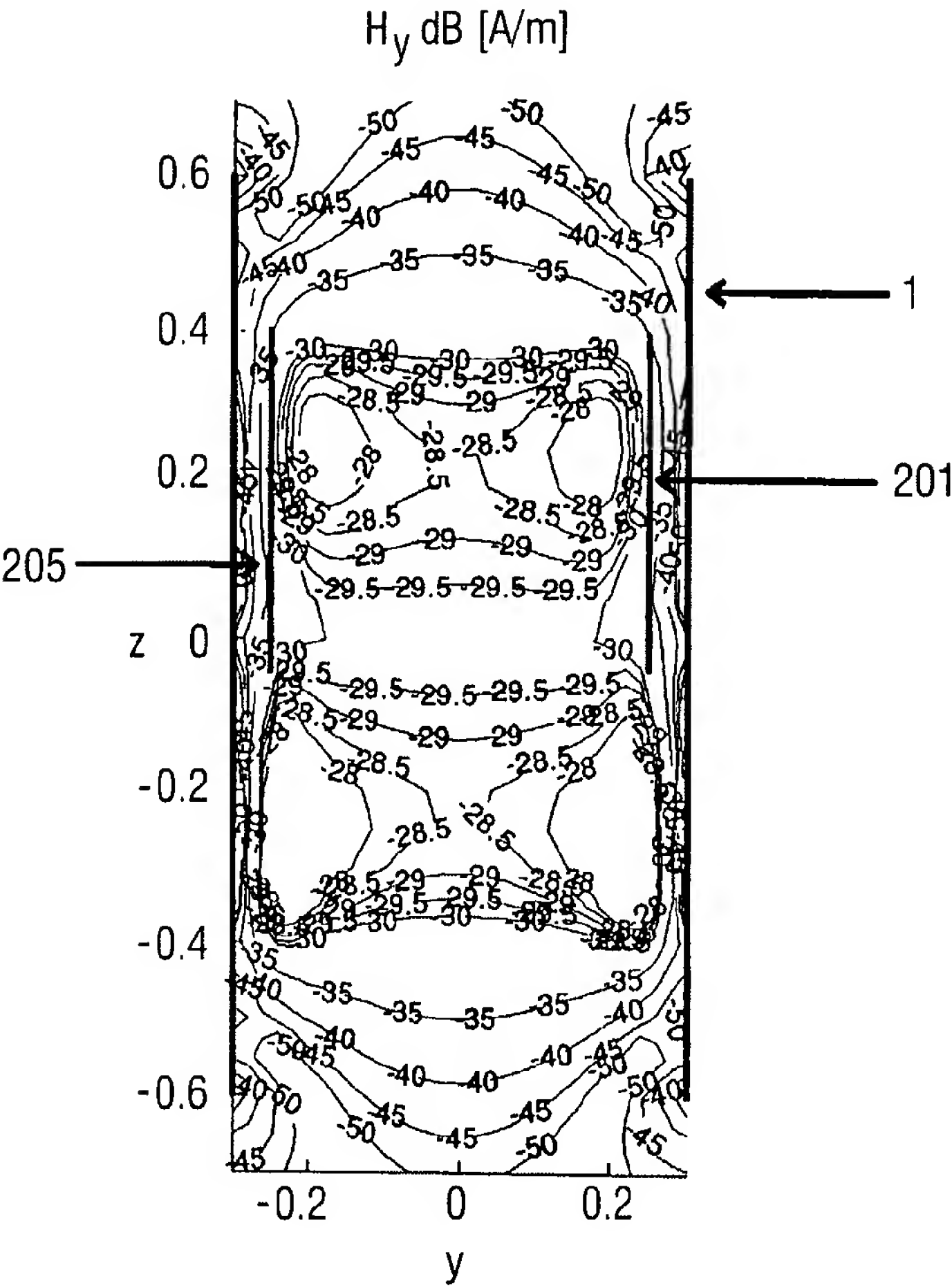


FIG.6

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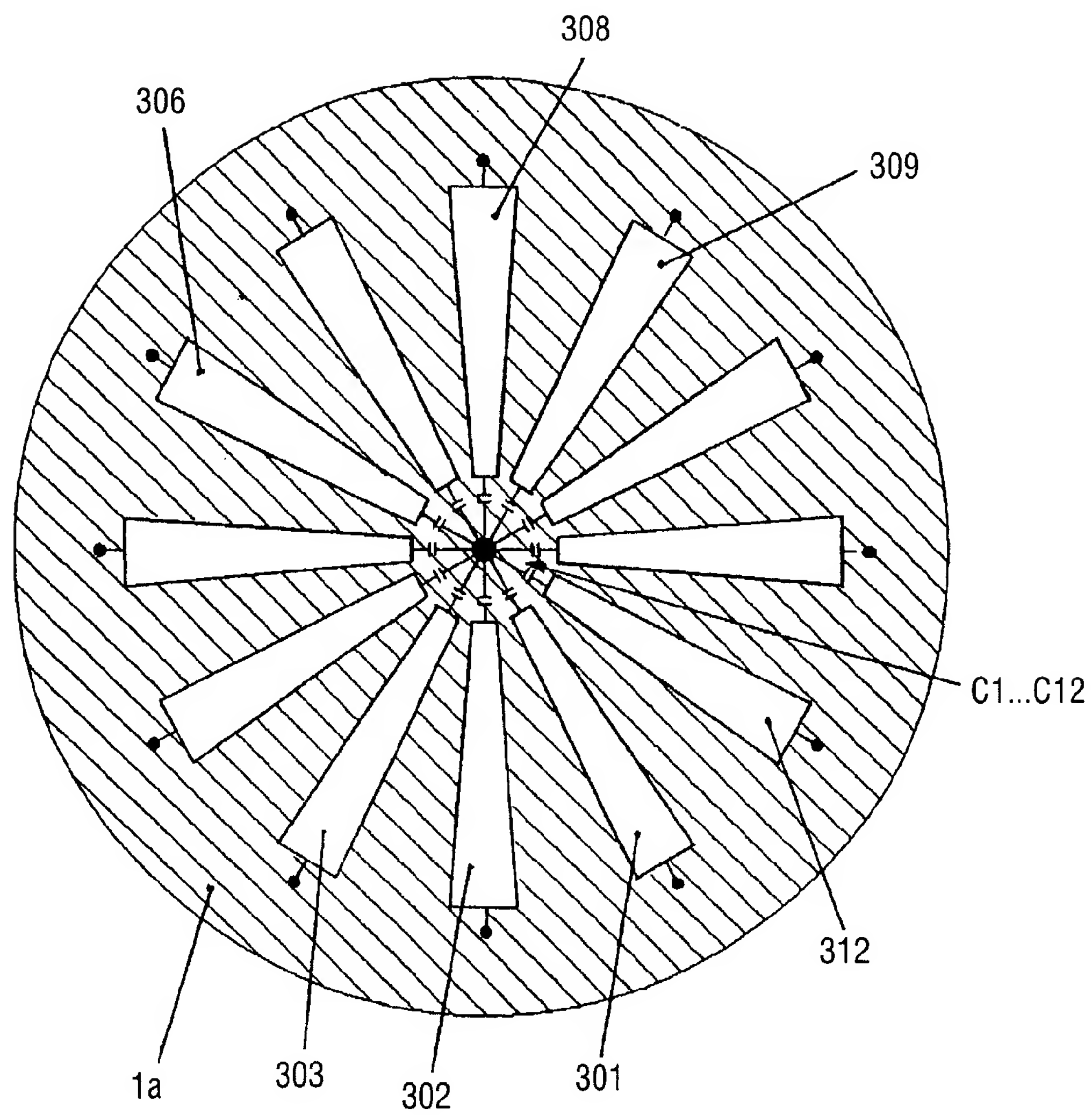


FIG. 7

INTERNATIONAL SEARCH REPORT

PCT/IB 03/05213

A. CLASSIFICATION OF SUBJECT MATTER

IPC 7 G01R33/34 G01R33/341

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

IPC 7 G01R

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

INSPEC, EPO-Internal, WPI Data, PAJ

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category *	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	LEE R F ET AL: "Planar strip array (PSA) for MRI" MAGNETIC RESONANCE IN MEDICINE, APRIL 2001, WILEY, USA, vol. 45, no. 4, pages 673-683, XP002267960 ISSN: 0740-3194 see the whole document --- -/--	1-3,5, 7-10



Further documents are listed in the continuation of box C.



Patent family members are listed in annex.

* Special categories of cited documents:

- *A* document defining the general state of the art which is not considered to be of particular relevance
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INTERNATIONAL SEARCH REPORT

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C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	DUERR W ET AL: "WELLENLEITERANTENNEN FUER DIE KERNSPINTOMOGRAFIE. \TEIL 2: ANTENNENOPTIMIERUNG" ARCHIV FUR ELEKTRONIK UND UBERTRAGUNGSTECHNIK, S.HIRZEL VERLAG. STUTTGART, DE, vol. 44, no. 4, 1 July 1990 (1990-07-01), pages 336-343, XP000147845 ISSN: 0001-1096 see chapter '3. Massnahmen zur Erzielung hoher Signal-Rauschverhältnisse (SNR)' ---	1-3,5,6, 8,9
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